RF COIL SYSTEM FOR SUPER HIGH FIELD (SHF) MRI

DESCRIPTION

This application relates to the magnetic resonance arts. It finds particular application in super high field magnetic resonance imaging (MRI), and will be described with particular reference thereto. However, it also finds application in other magnetic resonance applications such as magnetic resonance spectroscopy, low field imaging, and

5   the like.

Heretofore, magnetic resonance imaging scanners have included a main magnet, typically superconducting, which generates a spatially and temporally constant magnetic field B_0 through an examination region. A radio frequency (RF) coil, such as a whole-body coil, a head coil, and the like, and a transmitter have been tuned to the

10   resonance frequency of the dipoles to be imaged in the B_0 field. The coil and transmitter have often been used to excite and manipulate these dipoles. Spatial information has been encoded by driving the gradient coils with currents to create magnetic field gradients in addition to the B_0 field across the examination region in various directions. Magnetic resonance signals have been acquired by the same or separate receive-only RF coil,

15   demodulated, filtered and sampled by an RF receiver and finally reconstructed into an image on some dedicated or general-purpose hardware.

In MRI, RF coils are used to generate B_1 magnetic fields for exciting the nuclear spins and detecting signals from them. Two types of RF coils, birdcage and transverse electromagnetic (TEM) coils, are widely used for imaging. The birdcage coil

20   has multiple conductor rungs which are arranged around the examination region extending parallel to the main field direction. The parallel conductor rungs are connected to each other via an end cap or ring at one end of the coil and a circular loop conductor at the other end. Current flows back and forth through the rungs, the end cap, and the loop. The resonance behavior of the birdcage is determined by capacitor elements arranged

25   within the parallel conductors and the circular loop conductor. Birdcage coils exhibit a substantially uniform magnetic field distribution in their interior at frequencies at or under 128MHz, which correspond to proton imaging in a main B_0 magnetic field of 3T. However, for super high field applications (B_0 >3 T), the application of birdcage coils is limited with respect to radiation losses due to propagation effects inside the bore of the

MR system and strong loading effects of the tissue.   Typically, the losses become unacceptable when half the wavelength at resonance is less than the bore diameter.

The problem of radiation losses can be overcome by reducing the diameter of the RF bore. The cut-off frequency of a bore with traditional 0.7m diameter is 259 MHz. Hence, an open birdcage cannot be used for whole body MRI at 7T, where the proton resonance is 298 MHz and the half wavelength is even smaller. Reducing the diameter of the bore increases the cut off frequency, but the strong coupling to the tissue due to RF eddy currents ($\sim \omega^2$) is still a fundamental problem. The induced impedance in the conductors caused by the asymmetric subject loading can generate strong B_1 inhomogeneity. Maintaining patient loading symmetry becomes more difficult as the RF coil becomes smaller and approaches the subject more closely.

Similar to the birdcage coils, TEM coils include parallel conductors, which are arranged around the examination region. The TEM coil is typically open on both ends, lacking both the end cap and the circular loop conductor. The TEM coil typically includes an outer RF current return path. TEM coils tend to provide improved radio frequency performance compared with the birdcage coils for higher frequency applications ($B_0 > 3T$) due to lower radiation losses. Yet, the TEM coils have some disadvantages. For example, the TEM coils are leaky from the standpoint of electromagnetic fields, generating stray RF fields. The birdcage coil has better B_1 homogeneity in the axial z-direction.

There is a need for the super high frequency (SHF) coil system which provides good RF properties in high frequencies applications. The present invention provides an improved imaging apparatus and an improved method, which overcome the above-referenced problems and others.

In accordance with one aspect of the present invention, a radio frequency coil system for magnetic resonance imaging is disclosed. The coil system includes a plurality of parallel spaced apart rungs which each includes rung capacitors; an end cap disposed at a closed end of the coil system; and an RF shield which is connected to the end cap and surrounds the rungs extending in a direction substantially parallel to rungs.

In accordance with another aspect of the present invention, a method of using the coil system is disclosed. The coil system includes a plurality of parallel spaced apart rungs which each includes rung capacitors; an end cap disposed at a closed end of the coil system; and an RF shield which is connected to the end cap and surrounds the rungs

5   extending in a direction substantially parallel to rungs.

One advantage of the present invention resides in providing a uniform field when using a birdcage-like structure at super high frequencies.

Another advantage resides in its ability to function as a bandpass transmit/receive coil array.

10   Another advantage resides in its scalability.

Another advantage is that it has features of TEM and birdcage coils.

Another advantage resides in its ability to be configured to provide a double resonant coil system.

Still further advantages of the present invention will be appreciated to those

15,   of ordinary skill in the art upon reading and understand the following detailed description.

The invention may take form in various components and arrangements of components, and in various steps and arrangements of steps. The drawings are only for

20   purposes of illustrating the preferred embodiments and are not to be construed as limiting the invention.

FIGURE 1 is a diagrammatic illustration of a magnetic resonance imaging system in accordance with the present invention;

FIGURE 2 is a diagrammatic illustration of an end section of the coil

25   system of FIGURE 1;

FIGURE 3 is a diagrammatic illustration of a side view of the coil system of FIGURE 1;

FIGURE 4 is a diagrammatic illustration of electrical connections of a shielded birdcage coil embodiment of the coil system of FIGURE 1;

30   --   FIGURE 5 is a diagrammatic illustration of electrical connections of a   ⁻ shielded birdcage coil embodiment of the coil system of FIGURE 1 with an open-end ring at top section;

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FIGURE 6 is a diagrammatic illustration of electrical connections of a resonator embodiment of the coil system of FIGURE 1 including separate meshes;

FIGURE 7 is a diagrammatic illustration of electrical connections of a resonator embodiment of the coil system of FIGURE 1 including individual meshes with

5   coupling and decoupling networks;

FIGURE 8 is a diagrammatic illustration of impedance transformers to control coupling/decoupling with the coil system of FIGURE 1;

FIGURE 9 is a diagrammatic illustration of electrical connections of a resonator embodiment of the coil system of FIGURE 1 including individual meshes

10   decoupled by overlap;

FIGURE 10 is a diagrammatic illustration of electrical connections of a bandpass birdcage coil embodiment of the coil system of FIGURE 1;

FIGURE 11 is a diagrammatic illustration of electrical connections of a double resonant RF coil resonator embodiment of the coil system of FIGURE 1;

15   FIGURE 12 is a diagrammatic illustration of electrical connections of a triple resonant RF coil resonator embodiment of the coil system of FIGURE 1;

FIGURE 13 is a diagrammatic illustration of electrical connections of a double resonant RF coil resonator embodiment of the coil system of FIGURE 1.

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With reference to FIGURE 1, a magnetic resonance imaging scanner 10 includes a housing 12 defining a generally cylindrical scanner bore 14 inside of which an associated imaging subject 16 is disposed. A main magnet or main magnetic field coils 20 are disposed inside the housing 12. The main magnetic field coils 20 are arranged in a

25   generally solenoidal configuration in the preferred high field embodiment to produce a main B_0 magnetic field directed along a z-direction lying parallel to a central axis 22 of the scanner bore 14. The main magnetic field coils 20 are typically superconducting coils disposed inside in cryoshrouding 24, although resistive main magnets can also be used.

The housing 12 also houses or supports magnetic field gradient coils 30 for

30   selectively producing magnetic field gradients along the z-direction, along in-plane directions transverse to the z-direction (such as along Cartesian x- and y-directions), or along other selected directions. The housing 12 also houses or supports a high frequency

whole body RF coil 32 for selectively exciting and/or detecting magnetic resonances. The housing 12 typically includes a cosmetic inner liner 34 defining the scanner bore 14. In addition to the whole body radio frequency coil 32, a high frequency head RF coil system 38, which includes one or more RF coils 40, is disposed. As described in detail below, the

5   high frequency whole body and head RF coils have birdcage, TEM, hybrid, combination birdcage and TEM, or other embodiments. The coil system 38 is preferably circularly cylindrical, but, of course, might have other geometries, such as an elliptic cross-section, semi-circular cross-section, semi-elliptical cross-section, and the like.

The main magnetic field coils 20 produce a main magnetic field B_0 in a

10   z-direction which is preferably greater than 3.0T, such as 7.0T. A sequence controller 42 operates gradient controller 44 to selectively energize the magnetic field gradient coils 30 and operates a radio frequency transmitter 46 coupled to one or more of the radio frequency coils 32, 40 to selectively energize the radio frequency coil or coils 32, 40. By selectively operating the magnetic field gradient coils 30 and the one or more radio

15   frequency coils 32, 40, magnetic resonance is generated and spatially encoded in at least a portion of a selected region of interest of the imaging subject 16. A radio frequency receiver 48, or in a SENSE mode a receiver for each SENSE channel, is coupled to one or more of the radio frequency coils 32, 40 to receive encoded magnetic resonance k-space samples that are stored in a k-space memory 50.

20   A reconstruction processor 52 applies a suitable reconstruction algorithm such as an inverse Fourier transform reconstruction algorithm, a SENSE algorithm, or the like to reconstruct the k-space samples into a reconstructed volume image representation including at least a portion of the region of interest of the imaging subject. The reconstructed image representation is stored electronically in an image memory 54,

25   displayed on a user interface 56, stored in non-volatile memory, transmitted over a local intranet or the Internet, or otherwise viewed, stored, manipulated, or so forth. The user interface 56 can also enable a radiologist, technician, or other operator of the magnetic resonance imaging scanner 10 to communicate with the sequence controller 42 to select, modify, and execute magnetic resonance imaging sequences.

30   With reference to FIGURES 2 and 3, the RF coil 40 includes a plurality of spaced apart parallel conductors or rungs or strips 60 which extend longitudinally in the z-direction parallel to the main magnetic field B_0. The rungs 60 are made of printed

circuit board, copper foil, conductive rod, and the like, and, in one embodiment, include sections varying in size and width. The rungs **60** are shown on-end in phantom in FIGURE 2. Local RF magnetic field detectors, such as loop sensors (not shown) are connected with the coil to measure coil currents. A generally cylindrical radio frequency

5   screen or shield **62** surrounds the strips **60**. To minimize eddy currents, the RF screen **62** is preferably slotted foil, mesh, or the like. A conductive end cap **64** is disposed at an end section **66** of the coil system **38**. In FIGURE 2, Cartesian x-y-z coordinates are indicated corresponding to the spatial orientation of the coil system **38** in the cylindrical scanner bore **14** shown in FIGURE 1. However, it will be appreciated that the coil system **38** may

10   be used in other orientations.

With reference to FIGURE 4, in a birdcage embodiment of the coil **40**, each individual strip **60** includes a plurality of strip resonance frequency tuning capacitors **68** and is connected to the end cap **64**, which defines return paths for the rungs. The capacitors are preferably lumped and/or distributed PCB designs along the strips **60** to

15   prevent propagation effects. The end cap **64** is directly connected to the RF screen **62** which allows the birdcage coil **40** to work at super high frequency. Each strip **60** is connected to a copper ring **70**, which is disposed at a top section **72** of the coil system **38**, opposing the coil system end section **66**. Each of two neighboring strips **60** are coupled by end tuning capacitors **74** disposed along the ring **70** to define current paths. The coil

20   of FIGURE 4, which includes both the capacitors in the strips and in the ring, generally behaves as a band-pass coil. Depending on whether the strips **60** or segments of the ring **70** behave inductively, the bandpass coil **40** might behave as a high-pass or a low-pass type coil. The low-pass type is formed when the strip capacitors **68** of the strips **60** have much higher values than the end capacitors **74**, thus the segments of the end ring **70**

25   exhibit an inductive behavior; whereas, the strips **60** exhibit the capacitive behavior. The high-pass type is formed when the capacitors **68** of the strips **60** have much lower values than the end capacitors **74**, thus the segments of the end ring **70** exhibit the capacitive behavior; whereas, the strips **60** exhibit the inductive behavior. For the bandpass type, the capacitors **68**, **74** are proportioned so that neither a high-pass, nor a low-pass behavior

30   occurs.

With reference to FIGURE 5, the RF coil **40** is similar to the coil of FIGURE 4. The top section **72** is open. The individual strips **60** are directly connected to

the RF screen 62 (shown as a dotted line 76) at the top section 72. Each strip 60 is connected to a copper ring 78, which is disposed at the end section 66 of the coil system 38, opposing the coil system top section 72. Each of two neighboring strips 60 are coupled by closed end tuning capacitors 80 disposed along the ring 78 to define current paths. Such

5   configuration reduces RF components of the B_1 field component that is oriented in the z-direction, near the top section 72, and thus reduces SAR in this region of the coil.

With reference to FIGURES 6 and 7, the individual strips 60 are arranged into individual meshes 82 with each mesh having different phase reception characteristics. Each mesh 82 includes two strips 60 coupled through the end capacitors

10   74 connected in between the each two strips 60 at the open end section 72. The RF currents of the meshes 82 flow through individual end ring segments 84, which are not connected directly to the RF screen 62. The meshes 82 are coupled with the receiver 48 and, optionally, the transmitter 46, by couplings 86, which are preferably capacitive or inductive impedance circuits, to provide individual send/receive channels. In addition, by

15   the use of the impedance circuits between individual meshes, the mutual induction can be increased to provide higher mode separation and better robustness against asymmetric and heavy patient loading. In one embodiment, the couplings 86 include impedance transformers 88, shown in FIGURE 8. The impedance transformers 88 control the mutual magnetic coupling between the individual meshes 82. The couplings 86 can be adjusted

20   such that the coil 40 of FIGURES 6 and 7 can work as a volume coil generating a circularly polarized B_1 RF field, like a birdcage coil, or as a transmit/receive array, e.g., a SENSE coil. Such a transmit/receive array can generate a circularly polarized RF field and also can cover all mode patterns and generate individual RF field homogeneity required for RF shimming. Active switching components (not shown) are preferably used

25   to switch the capacitances (68, 74) to switch from the volume mode coil to the transmit/receive mode coil. Switching takes place without a need to exchange the coils.

With continuing reference to FIGURE 7, the individual meshes 82 are arranged into individual complete loops 90, which are not directly coupled to the end cap or RF shield. In addition to each two strips 60 being coupled through the open end

30   capacitors 74, the each two strips are also coupled through the closed end capacitors 80. In coil configuration of FIGURE 7, the RF screen 62 is not directly connected to the coil 40.

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With reference to FIGURE 9, another configuration of the decoupled meshes is shown, in which the individual meshes 82 are overlapped with each other, preferably to a degree that minimizes mutual inductive coupling.

With reference to FIGURE 10, the illustrated coil 40 is a bandpass birdcage
5   coil disposed inside an RF screen 62. Adjacent rungs 60 are coupled through the open end capacitors 74, disposed at the ring 70, and through the closed end capacitors 80. The ratio of capacitance of the capacitors 74, 68 is selected to decouple the current loops. The coil 40 can alternately be used as a transmit/receive array, in which each individual coil segment is connected to an individual transmit/receive channel, which each, preferably, has
10 fully controllable individual phase and amplitude. The coil system of FIGURE 10 provides geometric openness and excellent RF properties for high frequency applications.

With reference to FIGURE 11, the coil 40 is a double resonant RF coil resonator whose capacitances are selected such that it functions as a birdcage bandpass coil at the resonant frequency of one dipole and a TEM coil at the resonance frequency of
15 another dipole. Each rung 68 is connected to the end cap 64 and also to the end ring 70 at the open end 72. Each pair of rungs 60 is coupled by the open end capacitors 74, which are disposed along the end ring 70. Capacitors 92 are connected between the end ring 70 at a connection node 98 and the RF screen 62, creating a tunable current path through the RF screen 62. The current path through the capacitor 92 is tuned to the TEM resonance mode
20 at a higher resonance frequency; and to the rotating B_1 field bandpass mode at a lower frequency. For example, the TEM mode can be tuned to the proton (^{1}H) resonance frequency and the bandpass mode to the phosphorous (^{31}P) resonance frequency.

In one embodiment, shown in FIGURE 12, a triple resonant RF coil resonator is configured with lumped inductors 100 connected in parallel to the capacitors
25 92. The inductors 100 generate a third resonance mode.

In the double resonant RF coil resonator shown in FIGURE 13, the capacitors 92 are connected in between at least one point 102 along the rungs 60 and the RF screen 62 to provide a current path through the RF screen 62 and to tune the second resonance mode.

30 The invention has been described with reference to the preferred embodiments. Modifications and alterations may occur to others upon reading and understanding the preceding detailed description. It is intended that the invention be

constructed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.